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(71) Applicant(s)
Canon Research Centre Europe Ltd
(Incorporated in the United Kingdom)
20 Alan Turing Road, Surrey Research Park,
GUILDFORD, Surrey, GU2 5YF, United Kingdom

(72) Inventor(s)
Richard Anthony Kirk

(74) Agent and/or Address for Service
Beresford & Co
2-5 Warwick Court, High Holborn, LONDON,
WC1R 5DJ, United Kingdom

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H5R REN
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(54) Abstract Title
X-ray CCD detector having a second scintillator layer on back-thinned substrate

(57) An x-ray detector (1) provided for radiographic use comprising a back thinned substrate (3), a charge coupled device sensor array (5) which is sensitive to light provided at the front surface of the substrate (3), located at the front of the thinned portion of the substrate, and first and second scintillator layers. The first scintillator layer (19) is in optical contact with the sensor array (5) and responds to x-radiation by converting the x-radiation to light. The second scintillator layer (9) is in optical contact with the thinned portion of the substrate (3), at the back of the substrate and also responds to x-radiation by converting the x-radiation to light. Alternatively the detector may comprise an assembly of individual detectors (1,1',1") each comprised of a substrate, a charge coupled device sensor array (5) which is sensitive to light, provided at the front surface of the substrate and a scintillator layer (19) in optical contact with the sensor array. All of the individual detectors have a similar response to x-radiation and are vertically stacked in pixel registration.

FIG.1

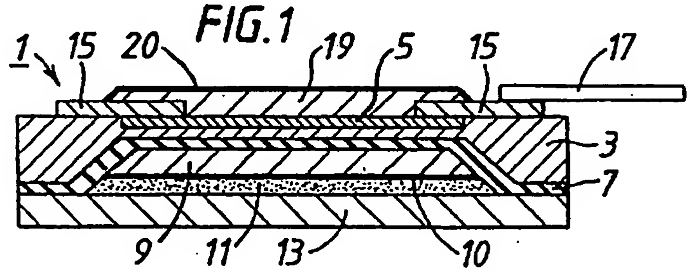

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FIG.1

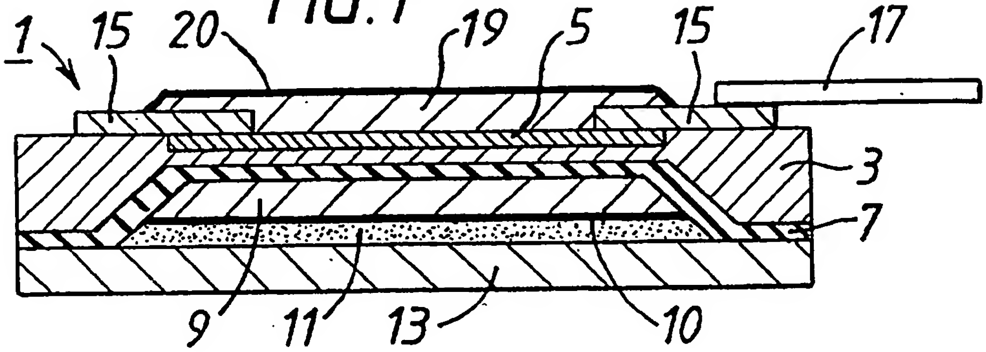


FIG.2

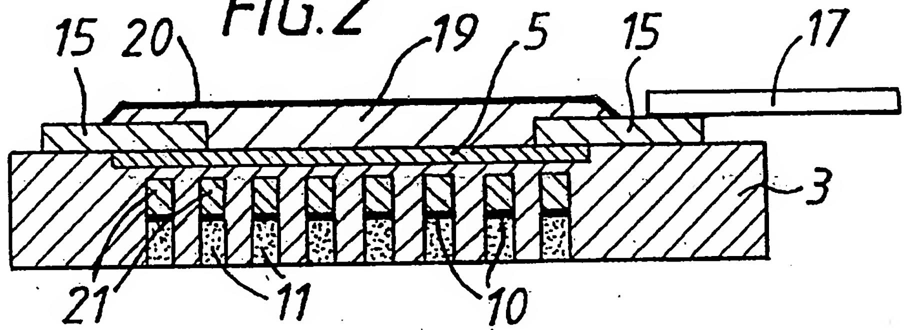


FIG.3

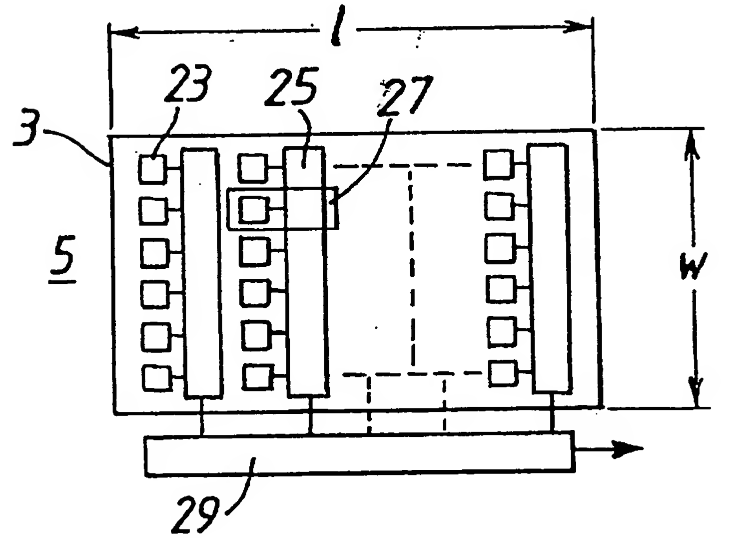


FIG. 4

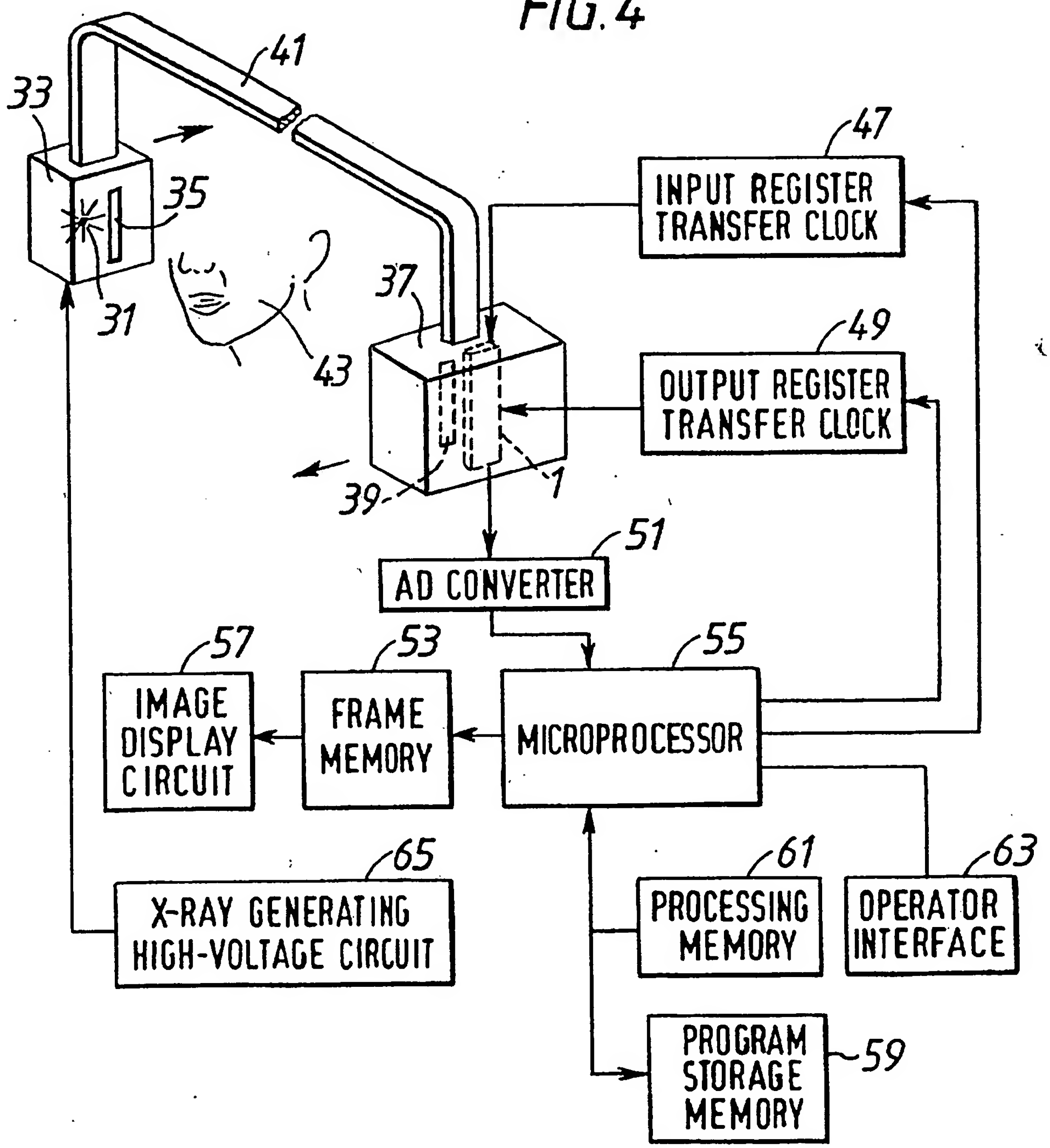
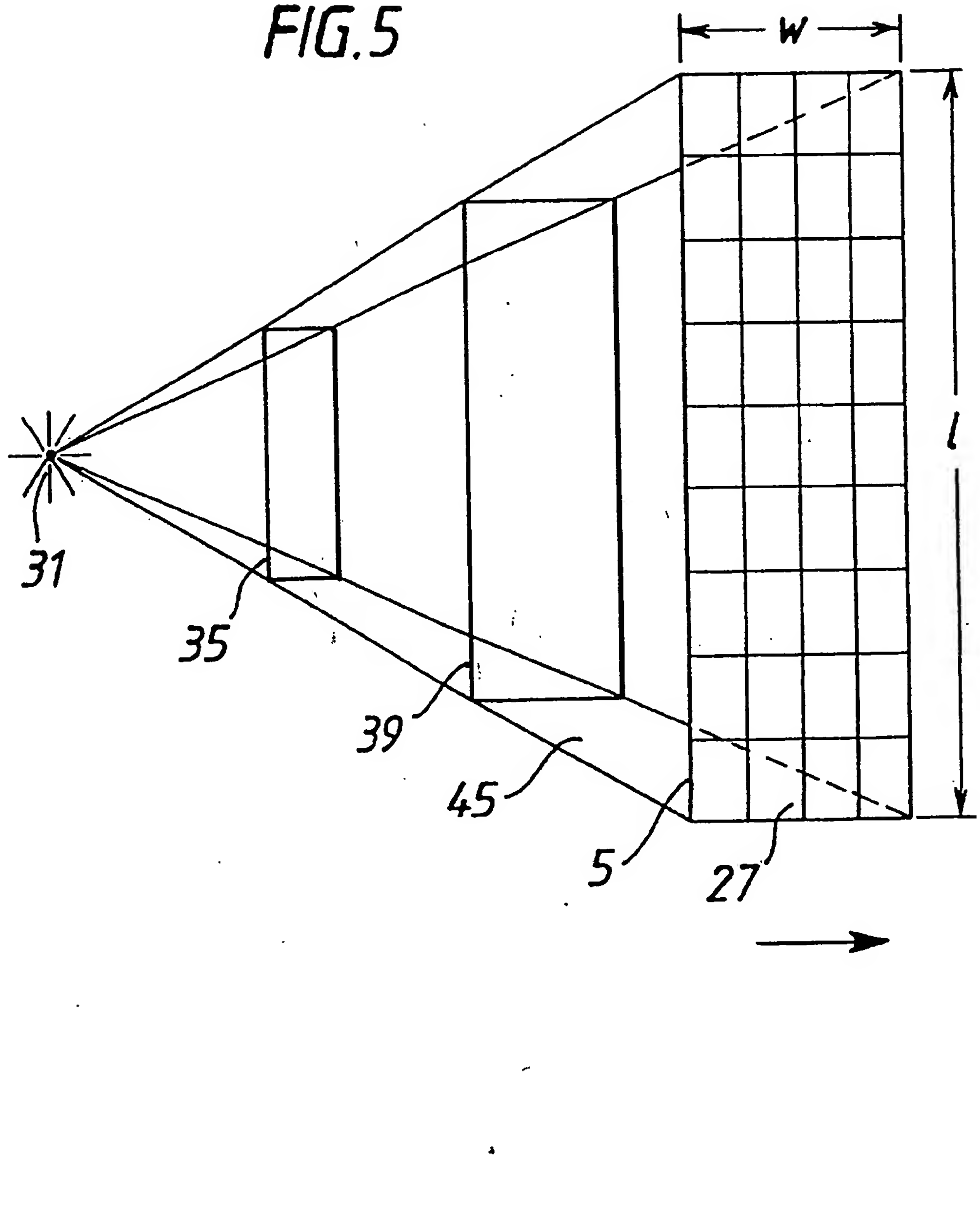

41
33
35
31
37
43
39
1
47
INPUT REGISTER TRANSFER CLOCK
49
OUTPUT REGISTER TRANSFER CLOCK
51
AD CONVERTER
57
IMAGE DISPLAY CIRCUIT
53
FRAME MEMORY
55
MICROPROCESSOR
65
X-RAY GENERATING HIGH-VOLTAGE CIRCUIT
61
PROCESSING MEMORY
63
OPERATOR INTERFACE
PROGRAM STORAGE MEMORY
59

FIG.5



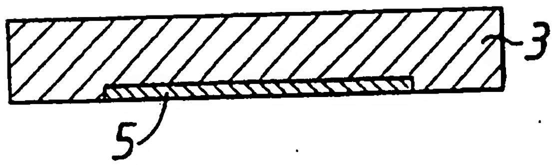


FIG.6(a)

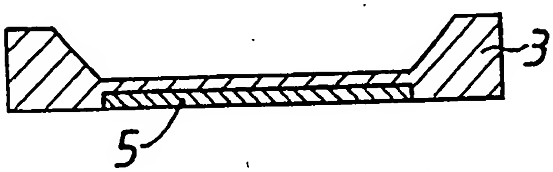


FIG.6(b)

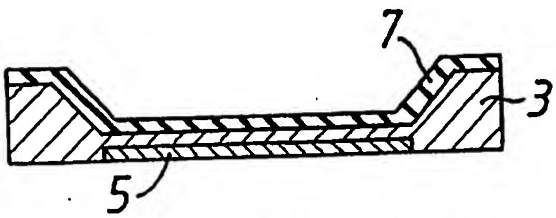


FIG.6(c)

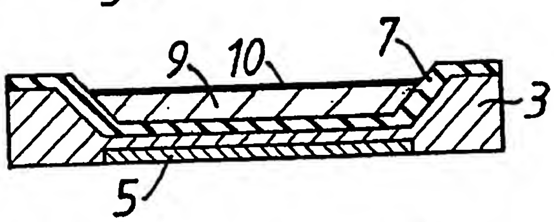


FIG.6(d)

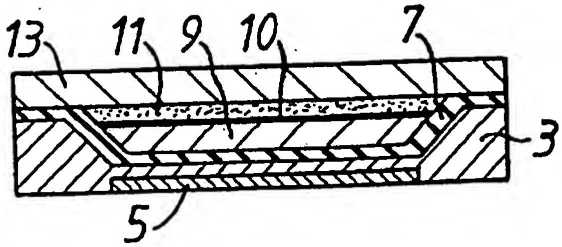


FIG.6(e)

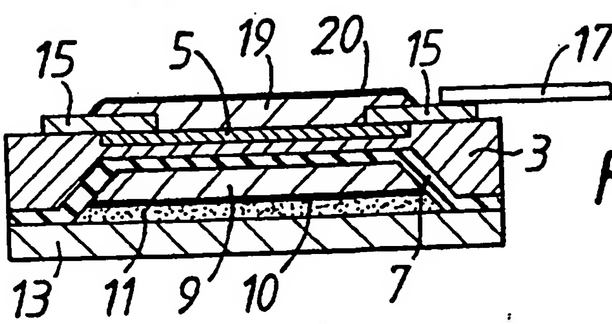


FIG.6(f)

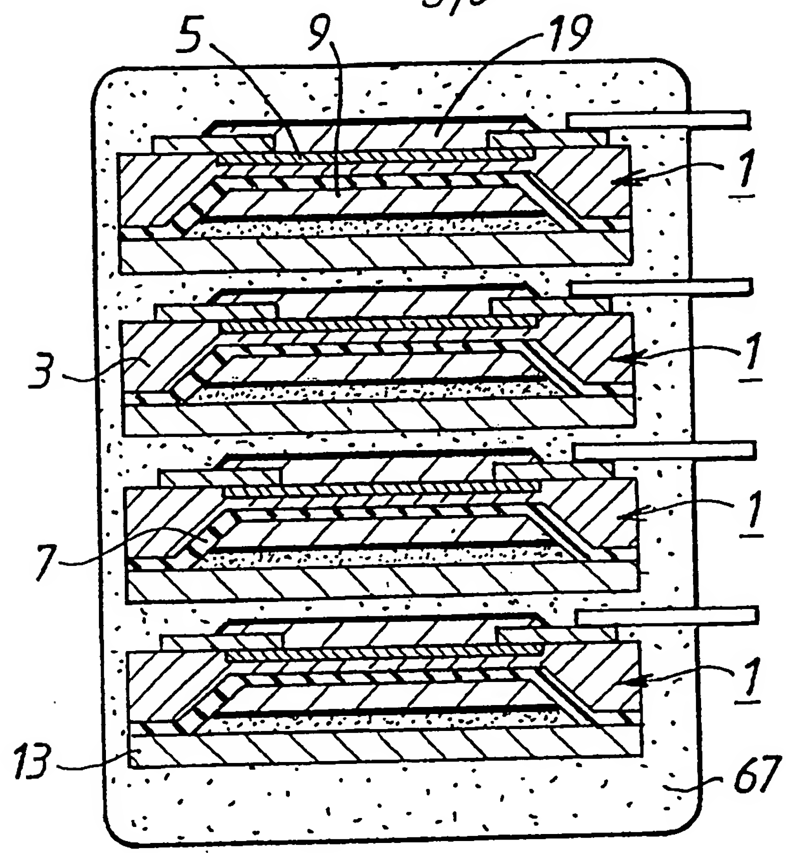


FIG.7

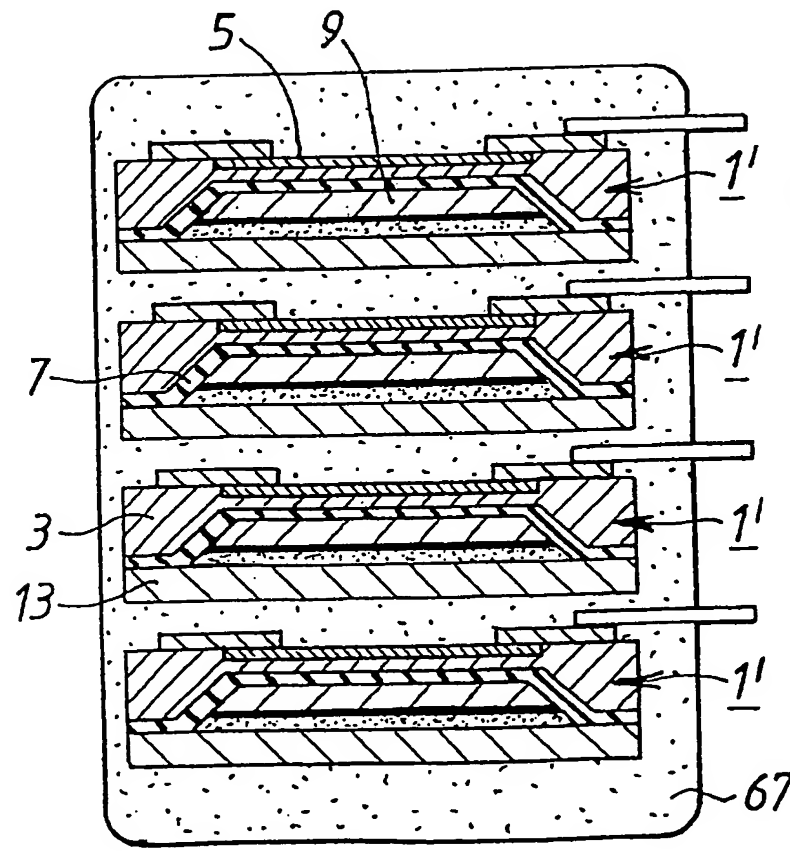


FIG.8

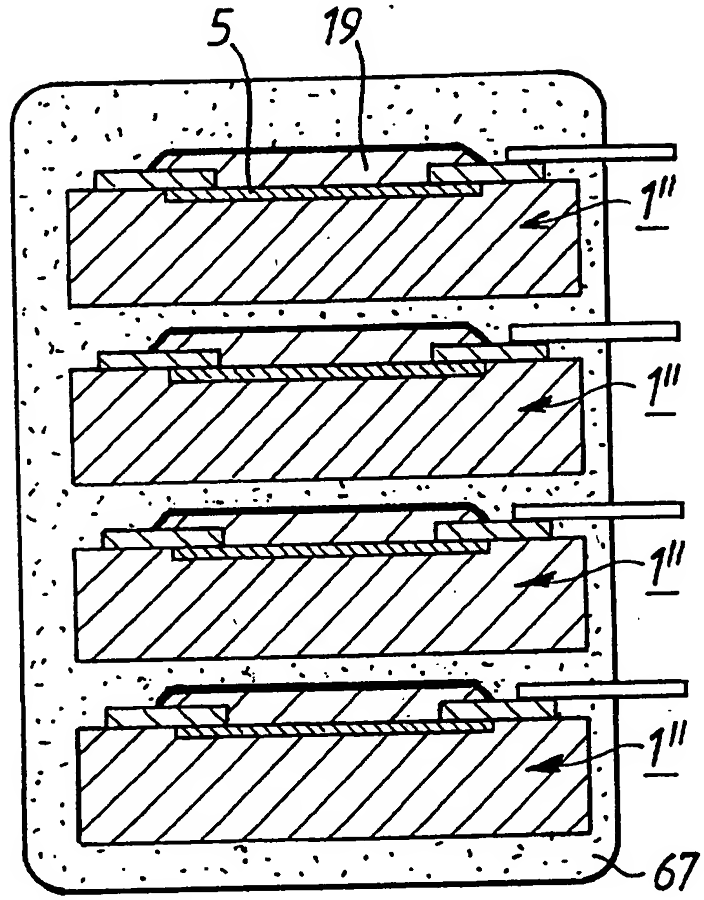


FIG. 9

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AN X-RAY DETECTOR FOR RADIOGRAPHIC USE

TECHNICAL FIELD

The present invention concerns X-ray detectors for radiographic use and, in particular, X-ray detectors for use in medical or dental examination.

When taking a radiograph, it is important to minimise the dose. This is because large doses and also high level accumulated dose have tendency to induce carcinogens. There is therefore a desire to optimise the efficiency of X-ray detectors for this use so that satisfactory detection and imaging can be performed using as small a dose as possible.

BACKGROUND ART

In case of X-ray detection, it is known to employ a detector which is sensitive to light, either UV or visible light, in combination with a screen of scintillator material which has the function of absorbing X-radiation and emitting light. Scintillator material is chosen as one which is effective for absorbing the source X-radiation and emitting light matched to the light sensitivity of the detector.

There is known an X-ray detector having a charge coupled device sensor array, sensitive to light, provided at the front surface of the substrate and having a scintillator layer in optical contact with the sensor array and which is responsive to X-radiation for converting the X-radiation to light. The scintillator

layer is in intimate optical contact with the sensor array in order to reduce optical cross coupling between the pixels of the sensor array. The thickness of the scintillator layer is chosen as a compromise between optimum X-radiation absorption efficiency and optimum resolution.

However, it is a problem with X-ray detectors of the kind just mentioned that polysilicon or metallised parts of the sensor array attenuate light radiation and, accordingly, reduce optical coupling efficiency.

As a means of overcoming this problem, there is known an X-ray detector comprising a back-thinned substrate, a charge couple device sensor array, sensitive to light, provided at the front surface of the substrate where it is located at the front of the thinned portion of the substrate and a scintillator layer which is in optical contact with the thinned portion of the substrate at the back surface of the substrate, and which is responsive to X-radiation for converting X-radiation to light. In this configuration, the polysilicon or metallised parts of the sensor array at the front of the detector are not located in the path of light emitted from the scintillator layer towards the sensor array.

SUMMARY OF THE INVENTION

The problem of providing further improvement in X-ray detector efficiency has been addressed and the present invention is intended as a means of solution to

that problem.

In the present invention, plural scintillator layers having a similar response to X-radiation are employed and means are provided for utilising the light converted from X-radiation by these scintillator layers for X-ray detection.

According to a first aspect of the present invention there is provided an X-ray detector for radiographic use comprising:

a back-thinned substrate;

a charge coupled device sensor array, sensitive to light, provided at the front surface of said substrate, located at the front of the thinned portion of the back-thinned substrate;

a first scintillator layer in optical contact with the sensor array and which is responsive to X-radiation for converting the X-radiation to light radiation; and

a second scintillator layer in optical contact with the thinned portion of the substrate, at the back surface of the substrate, similarly responsive to the same X-radiation, also for converting the X-radiation to light.

As the substrate, it is preferable to employ a substrate of monocrystalline semiconductor silicon, the active portions of the sensor array being defined in the surface of this substrate. Although alternative implementations using amorphous or polysilicon can be envisaged, these are less transparent than single crystal

silicon so less photons are likely to penetrate to the photosensitive regions of the detector.

It is also preferable that one or both of the first and second scintillator layers has a respective light reflective film coating at its outer surface for reflecting outwardly directed emitted light back towards the sensor array.

The back-thinned substrate may be one having a singular recess, in which case the second scintillator layer is a uniform continuous layer of scintillator material. In this case the void space of the recess may then be filled and/or may be covered by a rigid cover plate as means of increasing the rigidity, mechanical strength, and robustness of the detector. The filling, for example, might be of cured epoxy resin which material is chosen to have a thermal expansion coefficient matching that of the substrate.

Alternatively, the back-thinned substrate may be one having etch pits or trenches extending from its back surface. These pits or trenches are disposed at the back of the sensor active portions of the sensor array. In this case the second scintillator layer is not continuous but instead is composed of respective portions which are provided at the front bottom portion of each respective pit or trench. The substrate accordingly has greater rigidity and mechanical strength than in the case of a substrate having a singular recess. As before, the void

area is filled with a filler to provide a further enhancement in rigidity and mechanical strength. This likewise may be of a epoxy resin.

In the case of using monocrystalline semiconductor material, the length of the charge coupled device sensor array is limited by the length of sections that can be obtained from a monocrystalline wafer. In order to extend the length of these X-ray detectors, it is possible to provide a construction in which a plurality of detectors are arranged lengthwise in an abutted and bonded combination.

According to a second aspect of the present invention, there is provided an X-ray detector for radiographic use comprising:

an assembly of individual detectors each comprised of:

a substrate,

a charge coupled device sensor array, sensitive to light radiation, provided at the front surface of the substrate; and

at least one scintillator layer optically coupled to the sensor array for converting X-radiation to light; in which

the individual detectors have similar response to the same X-radiation and are vertically stacked in pixel registration.

It is preferable that each individual detector has

respective scintillator layers both front and back. Each individual detector may be an X-ray detector which is in accordance with the first aspect of this invention as described above.

However, the provision of individual detectors each having a scintillator layer either at its front or at its back is not precluded.

BRIEF DESCRIPTION OF THE DRAWINGS

Preferred embodiments of the present invention will now be described by way of example and particular reference will be made to the following drawings:

Fig. 1 shows an illustrative cross-section of an X-ray detector as an embodiment of the first aspect of the present invention;

Fig. 2 shows an illustrative cross-section of a variant of the X-ray detector shown in Fig. 1;

Fig. 3 is an illustrative plan view of a CCD sensor array of a kind which may be employed in the X-ray detector of Fig. 1;

Fig. 4 is a schematic and block diagram of radiographic apparatus which, except for the provision of an X-ray detector embodying the present invention, is of conventional configuration;

Fig. 5 is a schematic perspective diagram to show the relationship of X-ray source, exit and entry apertures, and sensor array for a synchronously scanned fanned beam arrangement;

Figures 6(a) to (f) are illustrative cross-section views to show stages in the manufacture of the X-ray detector shown in Fig. 1;

Figs. 7 to 9 are illustrative cross-sections to show a preferred embodiment and alternative embodiments of an X-ray detector in accordance with a second aspect of the present invention.

DESCRIPTION OF PREFERRED AND OTHER EMBODIMENTS

A preferred embodiment of an X-ray detector in accordance with the first aspect of the present invention will now be described with reference to Figs. 1 and 3 of the drawings.

In Fig. 1 there is shown an X-ray detector 1 comprising a substrate 3 having a charge coupled device sensor array 5, sensitive to light, provided at its front surface. The substrate 3 has a singular recess extending from its back surface towards the sensor array 5. The sensor arry is located at the front of the thinned portion of this substrate 3. In this example, the substrate 3 is of monocrystalline semiconductor silicon and the active portions of the sensor array 5 are defined in the front surface of this substrate.

The process of back-thinning the substrate can result in surface damage. This surface damage can provide centres for trapping charge. As means of compensating such trapped charge the back surface region of the substrate is ion implanted and an insulator layer is

provided at the back surface of the back-thinned substrate 3. Since the substrate in this example is of monocrystalline silicon, it is convenient to use as insulator layer a thermally grown layer of silicon dioxide.

A uniform continuous layer 9 of scintillator material is provided in intimate optical contact with the back surface of the thinned portion of the back-thinned substrate 3, i.e. along the front, bottom part of the recess of the back-thinned substrate. The thickness of this scintillator layer is determined as a compromise between optimum X-radiation absorption and optimum detector resolution.

In order to enhance optical coupling between this scintillator layer and the detector array, a light radiation reflective film coating 10 is provided at the back surface of this scintillator layer 9 for reflecting backwardly directed light emitted by this scintillator layer 9 towards the sensor arrays.

For added rigidity and mechanical strength, the void space between the back surface of the coated scintillator layer 9 and the hindmost back surface of the insulator layered substrate 3,7 is filled with a filling material 11 and there is a rigid cover plate 13 bonded to the hindmost back surface of the substrate. The filling material and the material of the cover plate are preferably of light molecular weight material and thus

have little attenuation effect on the X-radiation that would pass through the substrate. These materials have respective thermal expansion coefficients which are matched to that of the substrate. The filling material 11, for example, may be of cured epoxy resin and the cover plate 13 may be of monocrystalline silicon.

The X-ray detector 1 is provided with contact pads 15 for connection to the sensor array and fly leads 17 which have been pressure bonded to these contact pads.

The description thus far is that of a conventional back-thinned charge coupled device sensor (BCCD).

The X-ray detector 1 shown in Fig. 1 is distinguished by a further layer 19 of scintillator material. This layer 19 is in intimate optical contact with the sensor array 5 and also is responsive to X-radiation for converting the X-radiation to light radiation. This too may be coated with a light radiation reflective coating 20.

Different scintillator materials exhibit different sensitivities to X-radiation of different wavelengths. The materials of the front and back scintillator layers 9 and 19 are chosen to have similar sensitivity to X-radiation and accordingly may be of the same scintillator material. The result, therefore, is that the X-ray detector response is a sum of responses to X-rays of the same region/s of the X-radiation spectrum and not the sum of responses to X-radiation from widely different regions

of the X-radiation spectrum.

In the variant of this embodiment shown in Fig. 2, the back-thinned substrate 3 has not a singular recess but a plurality of etch pits or trenches which are disposed at the back of the sensor active portions of the sensor array. The scintillator layer 9 at the back of the X-ray detector is not in this case a uniform continuous layer of scintillator material but instead is composed of respective portions 21 which are provided at the front, bottom portion of each respective pit or trench.

As in the first example described with reference to Fig. 1, the void space is filled with a rigid material such as cured epoxy resin. Given the added strength of this structure, the provision of a cover plate 13 is optional and no cover plate is shown in Fig. 2.

The sensor array 5 may be of conventional configuration as shown in Fig. 3 of the drawings. This comprises rows of active sensors 23, each row being arranged extending in the widthwise direction of the substrate 3, in this case a semiconductor silicon chip. These active sensors 23 may consist in gated photosensitive capacitor wells, photodiodes or phototransistors. Each row of active sensors is connected to a respective charge transfer device 25, for example a charge coupled device input register, which likewise extends in the direction of the widthwise extent

of the semiconductor chip 3. One pixel 27 is shown in this figure and corresponds to one unit cell of the regular pattern of the matrix array. Typically, the area of the active sensor is about 80% of the total pixel area.

The output of each of the charge transfer devices 25 is connected to the input of an output charge transfer device 29, e.g. a charge coupled device output register. This output charge transfer device may be implemented as part of the semiconductor chip 3, in which case it is arranged extending in the lengthwise direction "l" of the chip 2. The arrangement shown in Fig. 3 is of hybrid construction wherein the output charge transfer device is provided off chip.

The X-ray detector described may be incorporated in a medical X-ray imaging time delay integration (TDI) fan scanning apparatus of which an example is shown in Fig. 4. It may of course, alternatively, be employed in different kinds of apparatus of different geometry to that shown.

The apparatus shown in Fig. 3 is, but for the incorporation of the X-ray detector 1, of conventional form such as that which is described in United States Patent US-A-5,664,001. This apparatus comprises an X-ray source 31 contained within a heavy metal enclosure 33 having a collimating exit aperture 35. The X-ray detector 1 is located within a heavy metal screen

enclosure which has a collimating entry aperture 39. A film (not shown) of metal, e.g. copper or aluminium, is put over the source target 31 and/or the exit aperture 39, in order to suppress soft X-rays. Enclosures 33 and 37 are mounted on a supporting frame member 41 such as a rotary arm swivel member as shown or, alternatively, a fixed yoke. The subject 43 of examination such as the jaw part of the head, in the case of dental examination, or chest or other body part, is interposed so as to lie in the path of the beam that is formed when X-radiation is emitted from the source through the exit aperture slit 35. The exit and entry apertures 35 and 39, which are vertical as shown in Fig. 4, are arranged in parallel.

The relationship between the source target 31, the exit and entry apertures 35 and 39, and the sensor array 5 of detector 1 is shown in Fig. 5. The divergent fan-out spread and width of the X-ray beam 45 are determined by the vertical height and slit width of the exit aperture 35 and the distance of this aperture from the X-ray source target 31. The fan-out spread and width of the beam illuminating the sensor array 5 of the detector 1 is determined by the distance of the sensor array 5 from the target 31 and the position and size of the entry aperture 39.

The arrow shown in Fig. 5 denotes the movement of the subject 43 relative to the array 5 that occurs during scanning. This direction is the direction of widthwise

extent of the sensor array 5 shown in Fig. 3 and corresponds to the transfer direction of each of the input registers 25. The timing of the transfer of charge along each input register is determined by an input register transfer clock 45. Charge transfer along the output register 25 is determined by an output register transfer clock 47.

The sensor array 5 is operated in a time delay integration (TDI) manner. Corresponding to movement of the subject 43, X-radiation emitted from the source target 31 through a given point of the subject 43 traverses the array of active sensors 23 pixel by pixel in the widthwise direction of the sensor array 5. On passing through each scintillator layer, the X-radiation is converted to light which illuminates the same pixel 23 and, as the subject 43 moves, each pixel 27 is illuminated in turn corresponding to the X-radiation passing through the aforesaid given point of the subject 43. The timing of the transfer of charge along each input register 25 is synchronised with the relative motion with respect to subject 43 so that, as charge is input from the active sensor 23 of each pixel 27 in turn, it is added to charge transferred from the previous input section of the input charge transfer register 25. In this way, charge is integrated in synchrony with movement of the subject 43 relative to the array 5 in the widthwise direction. The timing of the output register

transfer clock is chosen so that charge is input into the output register 29 from the output stage of each input register 25 and is transferred out of the output register 25 in the time of one input register transfer clock cycle.

As a typical example, a matrix sensor array having a pixel size 160 μm square will now be considered in the case of chest X-ray examination. The subject for X-ray is assumed to be 20 cm thick and it is also assumed that a resolution of least 0.2 mm for objects up to 20 cm from the detector is required. A typical small X-ray target spot size is 0.2 mm diameter and it is assumed that the exit aperture 35 is at a distance 20 cm from the target X-ray spot. The X-ray target is at a distance 1 m from the detector. There is a 5:1 ratio between the distance from the front of the chest to the detector and the distance from the X-ray spot to the detector, so the 0.2 mm spot will give a worst case geometric blur of 0.04 mm at the detector. If this is added to the stated pixel width of 0.16 mm, this will result in the desired resolution of 0.2 mm.

Since the exit aperture 35 is located 20 cm from the spot, this would have a 1:5 blur of the edges. The edge of the projected image will thus be 1 mm wider than the detector.

Supposing that the width of the exit aperture 35 is 1 mm, then for a point source the fan-out beam would have

a width at the detector of 5 mm and width 7 mm including X-ray radiation in the penumbra resulting from the blurring just discussed.

A TDI CCD sensor array 8 mm wide would thus be satisfactory to accommodate the width of the beam. This corresponds to 50 pixels each 160 μm square.

The length of the sensor array is determined by the number of such arrays that can be diced from a standard size wafer without significant wastage and thus at an economic cost.

In the case of chest X-ray examination, a width of about 50 cm at the detector is conventional. This may be accommodated by choosing six detector arrays arranged in line, each of length approximately 8 cm, each having 512 pixels per column of pixel size 160 μm square, 3K pixels per column in total. These may be diced from a single wafer of 8 inch diameter (20 cm) standard size.

Referring again to Fig. 4, the apparatus also includes an AD converter 51 for converting the analog signal output from the output register 29 to digital form which is then stored in a frame memory 53 under the control of a microprocessor 55. Data stored in the frame memory is then used to drive an image display via an image display circuit 57. The microprocessor 55 is part of a microcomputer which also includes a program storage memory 59 and a work space processing memory 61. Operator interaction is provided by an operator interface

63. An X-ray generating high voltage circuit 65 for controlling the X-ray source 31 is also shown.

A method of manufacturing the X-ray detector of Fig. 1 will now be described referring to Figs. 6(a) to (f). As shown in step (a) there is provided a monocrystalline silicon substrate 3 of thickness 0.3 mm having at its surface a CCD sensor array 5. The sensor array 5 has been produced in plural number in a single crystalline silicon wafer and the chip has been diced from this wafer.

In step (b) that part of the substrate 3 lying behind the sensor array 5 has been thinned down to about 20 μm thickness. This back-thinning is performed by chemical or plasma etching which may be performed in two stages, for example, an anisotropic etch to remove the bulk of the material and an isotropic etch to provide a flat defect-free smooth surface.

At step (c), ions are implanted to counter charge trapped at the etched surface and an insulator 7 of silicon dioxide is thermally grown under high pressure.

At step (d) a layer 9 of scintillator material is deposited over the thinned region of the substrate 3 by evaporation or as a powder with a binder. This makes good optical contact with the back surface of the silicon dioxide insulator layer 7. The thickness is matched to the desired resolution of the sensor array 5. A thin light reflecting layer 10 of aluminium is evaporated onto

the scintillator layer 9 to stop light escaping from the back of the detector 5.

At step (e), the remaining void behind the scintillator layer 9 is filled with epoxy resin and this is cured. A cover plate 13 of single crystal silicon is bonded at the back of the substrate 3. This is to provide further rigidity and mechanical strength.

At step (f), the device is flipped over and aluminium interconnects and contact pads are produced. This is followed by deposition of a second layer 19 of scintillator material which likewise may be produced by evaporation or using a powder mixed with binder. A thin light reflecting layer 20 of aluminium is evaporated onto the scintillator layer 9 to stop light escaping from the back of the detector. Lead wires 17 are ultrasonically bonded to the aluminium contact pads 15.

The detector produced at step (f) may then be packaged. For example, it may be set in epoxy resin.

In the case of manufacturing the variant shown in Fig. 2, step (b) is modified and etch pits or trenches are etched to correspond to the positions of the active sensor parts 23 of the sensor arrays. Ion implantation, thermal oxidation and scintillator layer deposition are performed as previously described at steps (c) and (d). Voids are filled with epoxy resin and cured as in the early stage of step (e). Step (f) is performed as above.

A preferred embodiment of an X-ray detector in

accordance with the second aspect of the present invention will now be described with reference to Fig. 7.

The detector shown comprises a plurality of individual X-ray detectors 1 each as described previously with reference to Fig. 1. These have respective sensor arrays 5 having the same array size and dimensions. The individual X-ray detectors are stacked vertically and are aligned so that the pixels of the respective sensor arrays are in register. The vertical stack of individual detectors 1 is encased in cured epoxy resin.

When arranged in use, the topmost detector will be closest to the subject and consequently the image produced for this individual detector will be sharper. However, provided that the total thickness of the stacked detector is below that of a few centimetres, then the image formed as the sum of images produced by each individual detector will not be degraded significantly for typical medical radiography.

The X-ray detectors of Fig. 1 discussed previously have each a thickness less than 0.5 mm. The thickness of a vertical stack of 20 such individual devices would be less than 1 cm. The parallax errors introduced over this distance would be negligible compared to errors resulting from examination of a subject 20 cm thick.

A vertical stack of twenty such individual detectors can be aligned without difficulty by placing them against a reference surface. Let us assume that each device is

1 cm wide and that the pixels are at least 100 μm across, e.g. 160 μm square as previously stated. A typical interference fit is about 2 μm. It would thus be possible to align each individual element to within 2 μm by placing them against a reference surface.

If the detectors have a skew misalignment so that the top and bottom individual devices are 50 μm out of registration with a central one of the individual detectors, this will not reduce noticeably the resolution given for mammography. This error corresponds to a rotation of $\tan^{-1}$ (0.01) or about half a degree. This would be easily visible and it can be corrected by eye.

X-ray detectors as described with reference to Fig. 1 and those described with reference to Fig. 7 may be arranged in line with other like detectors, abutted end to end and bonded together to form an elongate X-ray detector. This is similar to assembling optical line sensors, such as document readers the width of an A4 sheet, from a number of individual linear array sensors. Similar techniques of assembly can be adopted.

Alternative embodiments are shown in Figs. 8 and 9.

In Fig. 8 there is shown a vertical stack of individual detectors 1'. Each individual detector 1' is of the conventional back-thinned CCD sensor array type each having a scintillator layer 9 at its back surface only adjacent to the thinned portion of the substrate behind the sensor array.

In Figure 9 there is shown a stack of individual detectors 1". Each individual detector 1" is of the conventional type having the scintillator layer 19 at its front surface only adjacent the sensor arrays.

As to the support electronics for each of the examples described with reference to Figs. 6 to 8, this may be designed to sum the charges from the respective corresponding pixels before digitising. Alternatively, it is possible to digitise the charges prior to summation. In the latter case, the digitiser components may be integrated on each detector chip using spare area at the periphery of each chip.

The invention may be embodied in other specific forms without departing from the scope of the present invention. The present embodiments are therefore to be considered in all respects as illustrative and not restrictive, the scope of the invention being indicated only by the appended claims.

WHAT IS CLAIMED IS:

1. An X-ray detector for radiographic use comprising:

a back-thinned substrate;

a charge coupled device sensor array, sensitive to light, provided at the front surface of said substrate, located at the front of the thinned portion thereof;

a first scintillator layer in optical contact with said sensor array and which is responsive to X-radiation for converting the X-radiation to light; and

a second scintillator layer in optical contact with the thinned portion of said substrate, at the back surface of said substrate, similarly responsive to said X-radiation, also for converting the X-radiation to light.

2. An X-ray detector according to claim 1 wherein said substrate is of monocrystalline silicon, the active portions of said sensor array being defined therein.

3. An X-ray detector according to either preceding claim wherein said first scintillator layer has a light reflective film coating at its front surface for reflecting frontwardly directed light emitted by said first scintillator layer towards said sensor array.

4. An X-ray detector according to any preceding claim wherein said second scintillator layer has a light

reflective film coating at its back surface for reflecting backwardly directed light emitted by said second scintillator layer towards said sensor array.

5. An X-ray detector according to any preceding claim wherein the back surface of said substrate has an insulator layer interposed between the thinned portion of said substrate and said second scintillator layer.

6. An X-ray detector according to claim 5 wherein said insulator layer is of silicon dioxide.

7. An X-ray detector according to any preceding claim wherein said substrate has a singular recess extending from its hindmost back surface to the back surface of said thinned portion thereof, said second scintillator layer being a uniform continuous layer of scintillator material.

8. An X-ray detector according to claim 7 wherein the void space of the recess between the back surface of said second scintillator layer and the hindmost back surface of said substrate has a filling of light molecular weight material having a thermal expansion coefficient matching that of said substrate.

9. An X-ray detector according to claim 8 wherein said

filling is of cured epoxy resin.

10. An X-ray detector according to either of claims 7 or 8 having a reinforcement cover plate of light molecular weight material having a thermal expansion coefficient matching that of said substrate, located at its hindmost back surface.

11. An X-ray detector according to any of claims 1 to 6 wherein said substrate has a plurality of etch pits or trenches extending from its hindmost back surface to the back surface of thinned portions thereof, which pits or trenches are disposed at the back of sensor active portions of said sensor array, said second scintillator layer being composed of respective portions provided at the front bottom portion of each respective pit or trench.

12. An X-ray detector according to claim 11 wherein the void space of said pits or trenches, between the back surface of said respective portions of said second scintillator layer and the hindmost back surface of said substrate has a filling of light molecular weight material having a thermal expansion coefficient matching that of said substrate.

13. An X-ray detector according to claim 12 wherein said

filling is of cured epoxy resin.

14. An X-ray detector according to any preceding claim, encapsulated in cured epoxy resin.

15. An extended X-ray detector comprising a plurality of detectors as claimed in any preceding claim, which are arranged lengthwise in abutted and bonded combination.

16. An X-ray detector for radiographic use comprising:

an assembly of individual detectors each comprised of:

a substrate,

a charged coupled device sensor array, sensitive to light, provided at the front surface of said substrate; and

at least one scintillator layer optically coupled to said sensor array for converting X-radiation to light; wherein

said individual detectors have similar response to X-radiation and are vertically stacked in pixel registration.

17. An X-ray detector according to claim 16 wherein each individual detector has a scintillator layer in optical contact with its sensor array.

18. An X-ray detector according to claim 17 wherein said scintillator layer in optical contact with said sensor array has a light reflective film coating at its front surface for reflecting frontwardly directed light emitted by said scintillator layer towards said sensor array.

19. An X-ray detector according to any of claims 16 to 18 wherein each individual detector has a scintillator layer at its back, optically coupled to its sensor array.

20. An X-ray detector according to claim 19 wherein said scintillator layer at the back of the respective individual detector has a light reflective film coating at its back surface for reflecting backwardly directed light emitted by said scintillator layer towards said sensor array.

21. An X-ray detector according to claim 16 wherein each individual detector is an X-ray detector as claimed in any of claims 1 to 13.

22. An X-ray detector according to any of claims 16 to 21, encapsulated in cured epoxy resin.

23. An extended X-ray detector comprising a plurality of detectors as claimed in any of claims 16 to 22, which

are arranged lengthwise in abutted and bonded combination.

24. An X-ray detector constructed, adapted and arranged to perform substantially as described hereinbefore with reference to and as shown in any of Figures 1, 2 and 7 to 9 of the accompanying drawings.

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Application No:	GB 9912967.8	**Examiner:**	Eamonn Quirk
Claims searched:	1-15	**Date of search:**	28 October 1999

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Search Report under Section 17

Databases searched:

UK Patent Office collections, including GB, EP, WO & US patent specifications, in:

UK Cl (Ed.Q): H5R(RAD, REN, RTD) H1K (KEBA, KECCX)

Int Cl (Ed.6): G01T (1/00,1/20) H01L(27/148)

Other: Online: WPI, JAPIO, EPODOC

Documents considered to be relevant:

Category	Identity of document and relevant passage	Relevant to claims
A	GB 2 330 688 A (EEV Limited)	
A	GB 2 311 896 A (EEV Limited)	
A	EP 0 715 830 A1 (Picker International)	
A	US 5 398 275 (Catalin)	
A	US 5 715 292 (Loral Fairchild Corp.)	
A	US 4 798 958 (California Institute of Technology)	

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